

VARIABILITY OF PEAK SHOULDER FORCE DURING WHEELCHAIR PROPULSION
AS A FUNCTION OF SHOULDER PAIN

BY

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THESIS

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Abstract

Manual wheelchair users report a high prevalence of shoulder pain. Growing evidence shows that variability in forces applied to biological tissue is related to musculoskeletal pain. The purpose of this study was to examine the variability of forces acting on the shoulder during wheelchair propulsion as a function of shoulder pain. Twenty-nine manual wheelchair users (13 with pain, 16 without pain) participated in the investigation. Kinetic and kinematic data of wheelchair propulsion were recorded for three minutes maintaining a constant speed at three distinct propulsion speeds (fast speed of 1.1m/s, a self-selected speed, and a slow speed of 0.7m/s). Peak resultant shoulder forces in push phase were calculated using inverse dynamics. There was no difference in mean shoulder resultant force between groups. The pain group had significantly smaller variability of peak resultant force than the no pain group ($p < 0.01$). This highlights that manipulations of propulsion variability could potentially be a novel prevention strategy for upper limb pain in manual wheelchair users.

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1. Introduction

At least 2.6% (6.8 million) of the US population use assistive devices for mobility and nearly a quarter of those using assistive devices utilize a manual wheelchair for mobility [1]. Manual wheelchair users depend on their upper limbs for mobility and most functional activities. Unfortunately, the human upper limb is not specialized for the repetitive loading required for wheelchair propulsion. This requirement predisposes manual wheelchair users to upper limb pathology. Indeed, up to 70% of manual wheelchair users report upper limb pain [2], which is mainly manifested in the shoulder [3] and wrist [4]. Furthermore, even in manual wheelchair users who do not report pain, there is evidence of degenerative changes in the shoulder [5] suggesting that it is just a matter of time before these asymptomatic individuals will experience pain.

Upper limb pain in manual wheelchair users has been linked to difficulty performing activities of daily living, decreased physical activity and decreased quality of life [6]. Overall, any loss of upper limb function due to pain adversely impacts the independence and mobility of manual wheelchair users. It has been speculated that a decrease in independence and mobility results in greater health care costs and an increased risk for secondary morbidity (cardiovascular disease, obesity, etc.) [7]. Subsequently, it is imperative to understand the mechanisms that contribute to upper limb pathology in manual wheelchair users so that appropriate interventions can be developed to prevent or minimize the effect of pain on function and reduce the risk of long-term upper extremity disability. Factors that predispose manual wheelchair users to shoulder pain and pathology including demographics of manual wheelchair users and wheelchair configuration have been identified [8-14]. Research has observed that males, taller, heavier people and those with lower function are at greater risk for developing shoulder pain [8, 9].

An additional speculation is that propulsion mechanics (i.e. kinematic, kinetic and temporal components) contributes to the pathogenesis of shoulder pathology. This speculation is based on simple Newtonian mechanics that forces applied to the wheelchair hand rim resulted in reactive forces acting on the shoulder that may over time lead to musculoskeletal damage. For instance, an investigation demonstrated that greater tangential component on hand rim generates greater net moment around shoulder indicating higher risk of shoulder injury [14, 15]. Additionally, several investigations have analyzed force applied to the shoulder during wheelchair propulsion using inverse dynamics [5, 8, 9, 14-20]. These investigations have revealed that posterior force and superior force is the main force applied to the shoulder in the push phase [5, 8, 9, 14, 16]. The investigations discussed possible relationship between shoulder force and shoulder pathology suggesting that higher posterior force might be associated with coracoacromial ligament edema [5] while inferior force might cause compression of rotator cuff leading to shoulder impingement [5, 14, 20].

However, there has been a limited amount of research supporting a relation between forces acting on the shoulder during propulsion and shoulder pain or pathology [5, 8]. For instance, in 2006, Mercer et al. [5] investigated the relationship between shoulder force and shoulder pathology inferred through magnetic resonance imaging (MRI). It is important to note that although shoulder pathology was inferred from MRI, none of the participants self-reported shoulder pain at the time of assessment. In 2008, Collinger and colleagues [8] found that there was no difference in propulsion biomechanics between persons with and without pain. Consequently, there is a dearth of evidence that propulsion biomechanics are related to shoulder pain in manual wheelchair users.

A potential limitation of previous research is that it has almost exclusively focused on mean propulsion parameters. This approach essentially ignores the importance of movement variability as a factor related to musculoskeletal injury. It is maintained that movement

variability within limits is a normal characteristic of healthy neuromotor systems and affords greater adaptability to environmental stressors [21-24, 27]. There is growing evidence from experimental and clinical studies that motor variability is related to musculoskeletal injury [23-29].

Studies have reported that pain is related to decreased movement variability [21-30]. These investigations proposed that participants implemented an adaptive strategy of constrained motion to avoid pain [23, 24, 26, 28]. While pain has been shown to influence motor variability, evidence also has indicated that less variability may actually contribute to development of chronic pain [21, 29, 30]. These observations lead to the proposal that movements that are repeated more identically causes a significant amount of force applied to the same soft tissue in the same direction leading to chronic pain [30]. Therefore, it has been suggested that increased movement variability would modify forces applied to tissue from cycle to cycle distributing force equally among tissues [29, 30].

To date there has been minimal examinations of variability in manual wheelchair propulsion and its association to shoulder pain. The main goal of this investigation is to examine the relationship between shoulder pain and variability of peak force and moment on shoulder during wheelchair propulsion. It is hypothesized that variability of peak shoulder resultant force will be smaller in pain group than no pain group.

2. Methods

2.1. Participants

Twenty-nine manual wheelchair users (11 females, 18 males, age = 24.4 ± 9.7 years) who resided in the local community were recruited through advertisements. Inclusion criteria included (1) more than one year of manual wheelchair experience; (2) use manual wheelchair for 80% of their daily mobility; and (3) between 18-64 years of age. All subjects gave informed consent prior to participation to the study.

2.2. Demographics

Upon arrival to the laboratory participants were informed of the experimental procedures, provided the opportunity to ask questions about the procedures and then provided written informed consent. After providing informed consent, participants provided demographic information and self-reported shoulder pain. Specifically they answered a yes/no question concerning if they had current shoulder pain. Participants also completed the Wheelchair User's Shoulder Pain Index (WUSPI). The WUSPI was designed to measure the severity of shoulder pain related to functional activity of manual wheelchair users in daily living [31]. It is composed of 15 items relating to pain in everyday activities of wheelchair users including loading a wheelchair into a car, transferring etc. Participants reported their pain during each functional activity between zero and ten with a higher score indicating greater pain. Overall WUSPI score is sum of the 15 items with maximum possible score of 150. This measure has found to be valid and reliable in manual wheelchair users [31]. Participants were separated into groups based on their self-report of current shoulder pain (Pain = 13, No pain = 16).

2.3. Data collection and instrumentation

2.3.1. Anthropometric data

Based on previous research [8], segment length and upper extremity circumferences of all participants were measured as input to Hanavan's mathematical model [32] which calculates the inertial properties of each body segment used in inverse dynamic model. Body weight was measured by having participants wheel onto a force platform (AMTI, Inc., Watertown, MA) and then measuring the weight of the participant's empty wheelchair. Body weight was calculated as total weight minus wheelchair weight.

2.3.2. Kinetic data

Each subject's own wheelchair was fitted bilaterally with SMART wheels (Three Rivers Holdings, LLC, Mesa, AZ) and secured to a dynamometer system using a four-point tie-down system. The SMART wheel measures three-dimensional forces and torques applied to the push rim. Attaching the SMART Wheel to the subject's own wheelchair does not change the wheel placement, alignment, or camber [5]. All subjects were instructed to acclimate themselves to the dynamometer setup prior to testing. Kinetic data were collected at 100Hz and digitally filtered with an eighth-order, zero-phase, low-pass Butterworth filter with 10Hz cutoff frequency. Data were used after 30sec (first 3 cycles were not used), to give enough time to reached a steady-state speed.

2.3.3. Kinematic data

Motion capture system (Motion Analysis Co., Santa Rosa, CA) was used to collect kinematic data by tracking attached markers on the participant's body. Based on previous research [8], eighteen markers were attached bilaterally, at specific bony landmarks on the following location: third metacarpophalangeal joint (i.e. middle finger knuckle), radial styloid

(outside of wrists), ulnar styloid (inside of wrist), olecranon process (tip of elbow), lateral epicondyl, acromion (front of shoulder), sternal notch (chest), C7 vertebrae (base of neck), T3 vertebrae (base of skull), T6 vertebrae (middle region of the spine) and jaw. The kinematic data were collected at 100Hz and digitally filtered with fourth-order, zero-phase, low-pass Butterworth filter with 7Hz cutoff frequency.

2.3.4. Instrumentation

The participants were asked to propel at constant speeds of 4.0km/h (1.11m/s, fast), 2.5km/h (0.69m/s, slow), as well as a self-selected speed, for three minutes. The sequence of trial at different speeds was randomized for each subject. The self-selected speed was determined by asking the subject to push on the roller at a comfortable pace, as if they were pushing in a hallway. When the speed reached a steady state, the speed was recorded as the self-selected speed. Speed feedback was not given to the subject during this process. A speedometer was used to provide real-time feedback to the subjects.

2.4. Data analysis

2.4.1. Inverse dynamics

We used an inverse dynamic model described in an earlier investigation [33]. The model was used to compute 3-D joint forces and moments from segment kinematics, the forces acting on the push rim and subject anthropometrics.

Custom MATLAB code (The MathWorks, Natick, MA) were used to calculate shoulder joint forces using the inverse dynamic analysis. For participants with pain, the shoulder with the most pain was analyzed (Right = 10, Left = 3). Since the majority of participants with pain (79%) reported most pain in their dominant shoulder, the dominant shoulder was analyzed for participants without pain (Right = 12, Left = 4).

Shoulder resultant force was calculated based on 3-D shoulder joint force result from inverse dynamic analysis. Using custom MATLAB code, peak values of shoulder resultant force in push phase of each cycle was identified.

2.4.2. Definition of push, and recovery phase

Each cycle was defined to consist of push and recovery phases. The onset of push phase was defined as the point at which a propulsive moment, measured by the SMART wheel, was applied to the push rim, deviated from baseline by 5 percent [14, 34]. The end of push phase and the beginning of recovery phase were defined as the point when the propulsive moment measure by SMART wheel returned to baseline and remained within 5 percent [14, 34] (Figure 1) . Consistent with previous research, only the push phase of propulsion was analyzed [5, 8, 14, 17].

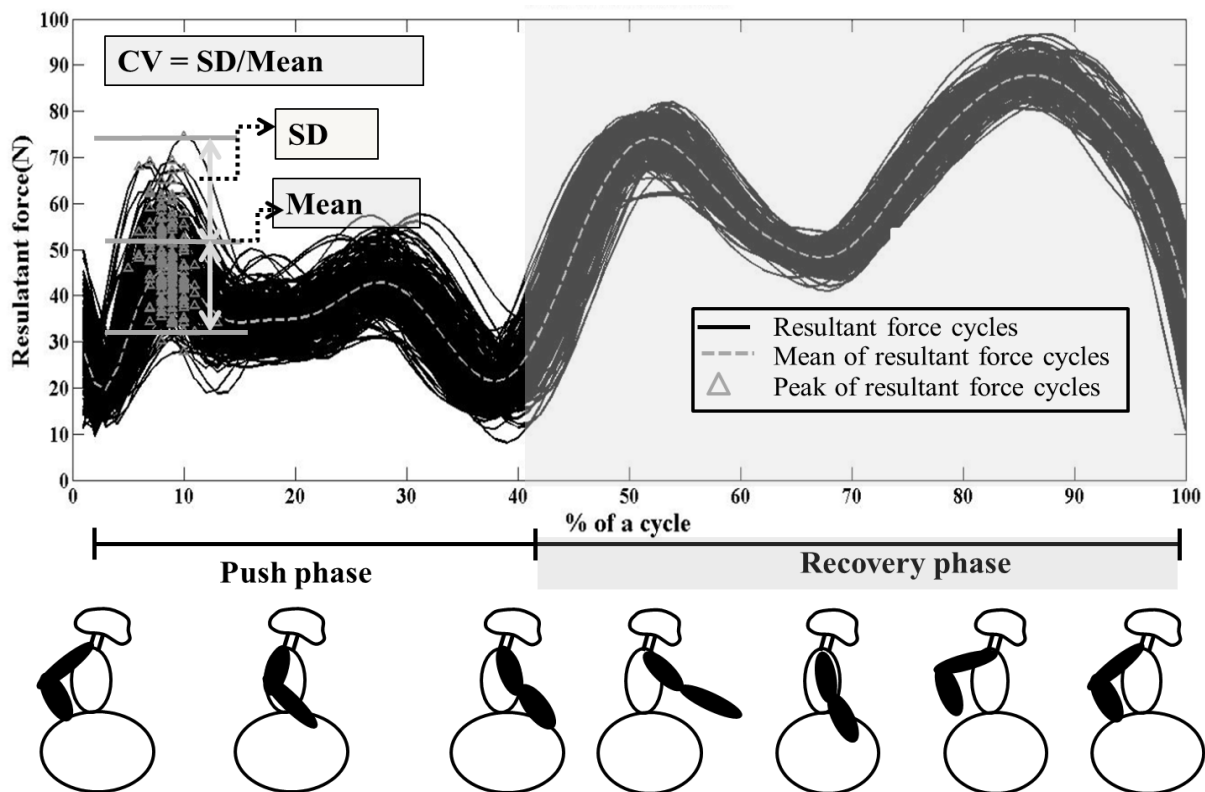


Figure 1. Cycle to cycle shoulder resultant force as a function of propulsion phase.

2.4.3. Variability of peak shoulder resultant force

For each individual participant, mean, standard deviation (SD), and coefficient variation ($CV=SD/Mean$) of peak values of shoulder resultant force in push phase over every stroke was calculated (Figure 1).

2.4.4. Statistical analysis

Statistical analysis was performed using SPSS for Windows, version 19.0 (IBM, Inc., Chicago, IL). Continuous demographic variables were analyzed with independent one-way ANOVAs and non-continuous variables were analyzed with independent-samples Mann-Whitney U Tests.

Individual mean, SD and CV of peak values of shoulder resultant force in push phase were entered into unique repeated measure analysis of covariance (ANCOVA) with weight as a covariant. Pain group (with and without) served as the between group factor while speed (slow, self and fast) served as the within subject factor. Weight was entered as a covariant since it has consistently been found to influence peak shoulder forces during wheelchair propulsion [5, 8, 14]. Significance (p-value) was set at ≤ 0.05 .

3. Results

3.1. Demographics

Demographic characteristics of the participants are reported in Table 1. The pain group had significantly greater shoulder pain than the no pain group as indicated by higher WUSPI scores [$F(1,27) = 12.5$, $p < 0.01$, $\eta^2 = 0.32$]. There was a significant age difference between groups [$F(1, 27) = 4.18$, $p = 0.05$, $\eta^2 = 0.13$], with the pain group being older than the no pain group. There was a trend for the pain group to be heavier than the no pain group, but traditional levels of significance were not reached [$F(1, 27) = 3.47$, $p = 0.07$, $\eta^2 = 0.11$]. There was no difference between groups in gender composition and the assessed shoulder. Spinal cord injury (pain = 38%, no pain = 50%) and spina bifida (pain = 38%, no pain = 25%) formed the majority of disabilities in both groups.

Table 1. Participant demographics as a function of pain group.

	N	Gender	Injury type	Shoulder side	Age (year)	Weight (kg)	WUSPI
Pain	13	M=7 F=6	Spinal cord = 5	R=10	28	75.6	22.8
			Spina bifida = 5 Amputation = 1 Spinal cyst = 1 Sacral Agenesis = 1	L=3	(13)	(25.4)	(21.3)
No pain	16	M=11 F=5	Spinal cord = 8	R=12	21	61.6	3.5
			Spina bifida = 4 Cerebral palsy = 2 Amputation = 1 Arthrogryposis = 1	L=4	(5)	(14.5)	(5.1)
p-value	--	0.42	--	0.91	0.05	0.07	<*0.01

Note: Values are mean (Standard deviation). M=Male, F=Female, R= Right, L= Left

3.2. Propulsion speed

Table 2. Propulsion speed as a function pain group and propulsion speed.

	Overall (m/s)	Pain (m/s)	No pain (m/s)	p-value
Slow speed (0.69 m/s)	0.73 (0.04)	0.73 (0.04)	0.73 (0.04)	0.77
Self selected speed	0.98 (0.20)	0.94 (0.15)	1.00 (0.24)	0.43
Fast speed (1.11 m/s)	1.14 (0.06)	1.13 (0.05)	1.15 (0.06)	0.29

Note: Values are mean (SD). p-value indicates group differences

There were no group differences in propulsion speed (p 's > 0.05). An analysis of propulsion speed revealed a main condition effect [$F(2, 84) = 81.58$, $p < 0.01$, $\eta^2 = 0.66$]. Post-hoc analysis revealed each speed condition was distinct.

3.3. Peak resultant force

Statistical analysis revealed that there was no significant difference in mean peak resultant force between two groups [$F(1, 26) = 0.14$, $p = 0.72$, $\eta^2 = 0.01$] (Figure 2). No significant effect of speed was found on mean peak shoulder force [$F(2, 52) = 2.07$, $p = 0.14$, $\eta^2 = 0.07$].

Figure 3 illustrates that SD was greater in the no pain group than the pain group. Descriptive analysis revealed that SD was 20%, 18% and 28% larger in the no pain group than the pain group in slow, self-selected and fast speed conditions respectively. Statistical analysis confirmed this observation [$F(1, 26) = 4.02$, $p = 0.05$, $\eta^2 = 0.13$]. No statistically significant effect of speed was observed [$F(2, 52) = 2.30$, $p = 0.11$, $\eta^2 = 0.08$].

Descriptive analysis showed that CV was 23%, 23% and 30% larger in the no pain group than the pain group in slow, self-selected and fast speed conditions respectively (Figure

4). Statistical analysis revealed that there was significantly larger CV in the no pain group than the pain group [$F(1, 26) = 22.9$, $p < 0.01$, $\eta^2 = 0.24$]. Statistical analysis showed that there was no speed effect on CV [$F(2, 52) = 1.81$, $p = 0.17$, $\eta^2 = 0.07$].

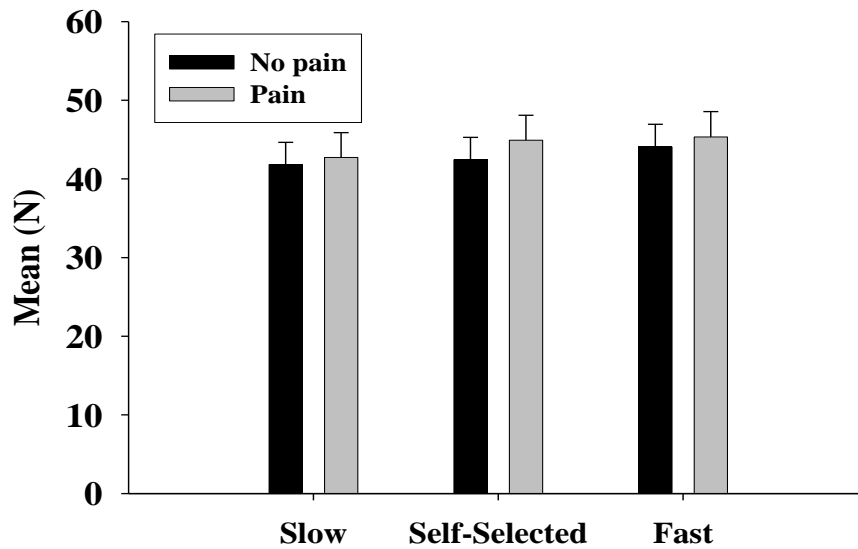


Figure 2. Mean peak shoulder resultant force as a function of pain group and propulsion speed.

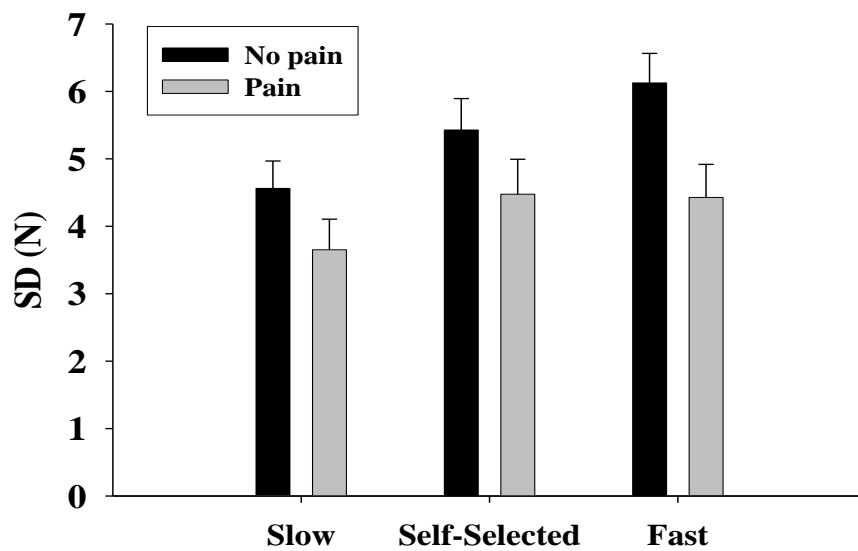


Figure 3. Standard deviation (SD) of peak shoulder resultant force as a function of pain group and propulsion speed.

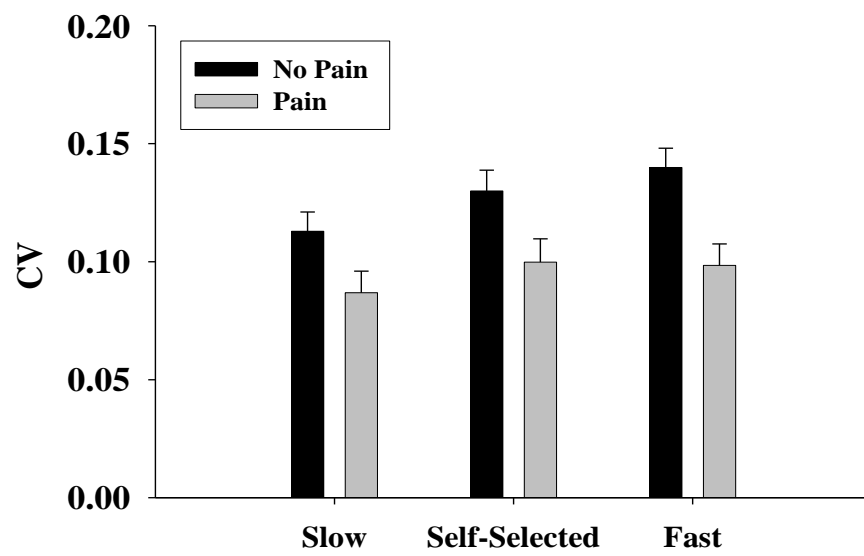


Figure 4. Coefficient of variation (CV) of peak shoulder resultant as a function of pain group and propulsion speed.

4. Discussion

The purpose of this investigation was to examine the relationship between shoulder pain in manual wheelchair users and variability of peak resultant force on the shoulder during wheelchair propulsion. Overall it was found that shoulder pain was associated with smaller variability of peak shoulder resultant force across a range of propulsion speeds. This study is one of the first demonstrations of an association between symptomatic shoulder pain and wheelchair propulsion biomechanics. The identification of the association between shoulder pain and kinetic variability highlights that manipulations of propulsion variability could potentially be a novel prevention strategy for upper limb pain in manual wheelchair users.

While shoulder pain is quite common in manual wheelchair users and propulsion mechanics are believed to play a significant role in the development of upper extremity pain [1], no previous research has documented an association between forces acting on the shoulder and symptomatic shoulder pain [5, 8]. It is important to note that these previous investigations have focused on mean parameters. Consistent with these studies, there was no difference in mean peak shoulder resultant force as a function of shoulder pain observed here. Yet, an examination of variability of resultant shoulder forces revealed significant difference between those with/without pain. The current observation highlighted that movement variability in of itself is sensitive marker of musculoskeletal pain in manual wheelchair uses [21].

Although this association between variability and self-reported pain is novel within wheelchair propulsion research, it is consistent with motor control/biomechanics research that has demonstrated that variability can play a functional role in the prevention or development of injury [21]. In general, movement variability stems from individuals adjusting their movements according to various dynamic constraints [21]. Movement variability has been

known to be associated with physiological adaptive process in maintaining tissue health. It has been suggested movement variability might play a positive role in preventing physiological injury by distributing force magnitude or direction acting on tissue to decrease rate of tissue damage. There are at least two potential reasons that the no pain group demonstrated greater variability in peak shoulder moment than the pain group [22]. First, it is possible that shoulder pain caused individuals to constrain their shoulder range of motion in order to avoid pain which resulted in decreased variability. Research has demonstrated that persisting pain is associated with decreased motor variability in a several repetitive motor tasks [23, 24, 26, 28]. It was suggested that participants were adapting movement solution that prevented pain. In contrast, experimentally induced acute pain was related to an increase in motor variability since people were searching for the movement strategy to reduce the pain [22]. Consistent with this data, the lower amount of variability in the shoulder pain group observed here might reflect participants constraining their motion to avoid pain.

Secondly, it is possible that lower amounts of variability of peak shoulder force could be an underlying mechanism that led to the development of shoulder pain by demanding relatively constant load on shoulder. In examinations of movement variability related to pain, the amount of forces applied to biological tissue can be divided into three categories: 1) normal healthy region where physiological force is mild enough to prevent pain; 2) chronic injury region where long term repeated performance within the region could cause chronic injury; and 3) acute injury region where force applied to tissue is over physiological limit of the tissue [21]. Obviously, the cross-sectional nature of this investigation cannot further comment on this possibility.

An unresolved question is where variability in shoulder kinetics stems from. Within the current investigation, an inverse dynamic model which depended on the force acting on the push rim and kinematic data of the upper limbs was used to determine net resultant shoulder

forces. Consequently, it is possible that the variability in resultant force comes from either fluctuations in forces at the hand rim, upper limb movements or a combination of the two. Preliminary data from our laboratory indicates lower variability of forces at the hand rim in persons self-reporting shoulder pain compared to those without shoulder pain during wheelchair propulsion [35]. Future research needs to examine the underlying contribution to variability in resultant shoulder forces is not only of theoretical importance but also has rehabilitative implications. Simply put the factors driving resultant force variability could be targeted for interventions. For instance, it is well known that wheelchair propulsion can be altered with various training interventions [36, 37]. It is within the realm of possibility that wheelchair users could be trained to move their arms in a more variable movement pattern [38]. In contrast, it is also possible to mechanically alter the variability of the force applied to rim. For example, it is likely that a flexible hand rim results in an increase in the variation in the forces applied to the rim.

Another novel aspect of this investigation was the collection of propulsion data over 3 minutes. On average this resulted in sampling of 130 strokes per a trial. Although the exact number of strokes required to fully characterize propulsion mechanics is not clear, this technique seemingly has advantages to procedures in which only a few strokes are sampled [5, 8, 14]. A caveat to this methodological practice is the increased possibility that participants became fatigue as the trials progressed. An examination of propulsion speed within in a given trial demonstrated no decline over time indicating that fatigue was minimal.

In addition, the study revealed that CV was more sensitive to shoulder pain than SD (e.g. had greater effect sizes (η^2)). This supports the notion that CV, as an index of relative variability is a more accurate parameter for estimating the effect of function on motor output. A limitation of SD is that it generally increases or decreases proportionally with the mean of the motor output [39].

Despite the novel observations of this current investigation, it is not without limitations. A major limitation of the study design is its cross-sectional nature. Consequently, this investigation cannot address whether decreased variability results from or contributes to the development of shoulder pain in manual wheelchair users. Further work may investigate the relationship between pain and variability of shoulder forces with a longitudinal design or an experimental model of shoulder of pain. Additionally, the generalizability of the current findings to the general wheelchair population is suspect – given that the current sample was primarily young college-age students. It is also important to note that the majority of our sample was child onset wheelchair users – which may impact the pathogenesis of musculoskeletal pain [40]. Also the current investigation focused on the amount of variability, this approach provides no information concerning the time sequential structure of the movement fluctuations [41]. Further research should investigate time sequential structure of variability in wheelchair propulsion. Lastly, while the current investigation examined resultant force applied to the shoulder, it is possible that the examination of uni-dimensional forces and moments could yield complimentary observations.

5. Conclusion

To our knowledge, this is one of the first investigations to identify an aspect of wheelchair propulsion related to symptomatic shoulder pain in manual wheelchair users. This observation supports the proposal that variability is a unique marker of movement that is related to musculoskeletal pain. Further, it raises the possibility that variability of wheelchair propulsion could be used as a clinical measure of shoulder pathology and manipulations of propulsion variability could potentially be a novel prevention strategy for upper limb pain in manual wheelchair users.

6. References

- [1] LaPlante, M. P., & Kaye, H. S. (2010). Demographics and trends in wheeled mobility equipment use and accessibility in the community. *Assistive Technology*, 22(1), 3-17.
- [2] Finley, M. A., Rasch, E. K., Keyser, R. E., & Rodgers, M. M. (2004). The biomechanics of wheelchair propulsion in individuals with and without upper-limb impairment. *Journal of Rehabilitation Research and Development*, 41(3 B), 385-394.
- [3] Curtis, K. A., Drysdale, G. A., Lanza, R. D., Kolber, M., Vitolo, R. S., & West, R. (1999). Shoulder pain in wheelchair users with tetraplegia and paraplegia. *Archives of Physical Medicine and Rehabilitation*, 80(4), 453-457.
- [4] Gellman, H., Chandler, D. R., Petrusek, J., Sie, I., Adkins, R., & Waters, R. L. (1988). Carpal tunnel syndrome in paraplegic patients. *Journal of Bone and Joint Surgery - Series A*, 70(4), 517-519.
- [5] Mercer, J. L., Boninger, M., Koontz, A., Ren, D., Dyson-Hudson, T., & Cooper, R. (2006). Shoulder joint kinetics and pathology in manual wheelchair users. *Clinical Biomechanics*, 21(8), 781-789.
- [6] Gutierrez, D. D., Thompson, L., Kemp, B., Mulroy, S. J., Winstein, C. J., Gordon, J., . . . Sullivan, K. (2007). The relationship of shoulder pain intensity to quality of life, physical activity, and community participation in persons with paraplegia. *Journal of Spinal Cord Medicine*, 30(3), 251-255.

- [7] Hardy, S. E., Kang, Y., Studenski, S. A., & Degenholtz, H. B. (2011). Ability to walk 1/4 mile predicts subsequent disability, mortality, and health care costs. *Journal of General Internal Medicine*, 26(2), 130-135.
- [8] Collinger, J. L., Boninger, M. L., Koontz, A. M., Price, R., Sisto, S. A., Tolerico, M. L., & Cooper, R. A. (2008). Shoulder biomechanics during the push phase of wheelchair propulsion: A multisite study of persons with paraplegia. *Archives of Physical Medicine and Rehabilitation*, 89(4), 667-676.
- [9] Gil-Agudo, A., Del Ama-Espinosa, A., Pérez-Rizo, E., Pérez-Nombela, S., & Pablo Rodríguez-Rodríguez, L. (2010). Upper limb joint kinetics during manual wheelchair propulsion in patients with different levels of spinal cord injury. *Journal of Biomechanics*, 43(13), 2508-2515.
- [10] Desroches, G., Aissaoui, R., & Bourbonnais, D. (2006). Effect of system tilt and seat-to-backrest angles on load sustained by shoulder during wheelchair propulsion. *Journal of Rehabilitation Research and Development*, 43(7), 871-882.
- [11] Veeger, H. E. J., Van Der Woude, L. H. V., & Rozendal, R. H. (1989). The effect of rear wheel camber in manual wheelchair propulsion. *Journal of Rehabilitation Research and Development*, 26(2), 37-46.
- [12] Gutierrez, D. D., Mulroy, S. J., Newsam, C. J., Gronley, J. K., & Perry, J. (2005). Effect of fore-aft seat position on shoulder demands during wheelchair propulsion: Part 2: An electromyographic analysis. *Journal of Spinal Cord Medicine*, 28(3), 222-229.

- [13] Rankin, J. W., Kwarciak, A. M., Richter, W. M., & Neptune, R. R. (2012). The influence of wheelchair propulsion technique on upper extremity muscle demand: A simulation study. *Clinical Biomechanics*, 27(9), 879-886. doi: 10.1016/j.clinbiomech.2012.07.002
- [14] Koontz, A. M., Cooper, R. A., Boninger, M. L., Souza, A. L., & Fay, B. T. (2002). Shoulder kinematics and kinetics during two speeds of wheelchair propulsion. *Journal of Rehabilitation Research and Development*, 39(6), 635-649.
- [15] Desroches, G., Aissaoui, R., & Bourbonnais, D. (2008). Relationship between resultant force at the pushrim and the net shoulder joint moments during manual wheelchair propulsion in elderly persons. *Archives of Physical Medicine and Rehabilitation*, 89(6), 1155-1161.
- [16] Van Drongelen, S., Van Der Woude, L. H., Janssen, T. W., Angenot, E. L., Chadwick, E. K., & Veeger, D. H. (2005). Mechanical load on the upper extremity during wheelchair activities. *Archives of Physical Medicine and Rehabilitation*, 86(6), 1214-1220.
- [17] Kulig, K., Newsam, C. J., Mulroy, S. J., Rao, S., Gronley, J. K., Bontrager, E. L., & Perry, J. (2001). The effect of level of spinal cord injury on shoulder joint kinetics during manual wheelchair propulsion. *Clinical Biomechanics*, 16(9), 744-751.
- [18] Robertson, R. N., Boninger, M. L., Cooper, R. A., & Shimada, S. D. (1996). Pushrim forces and joint kinetics during wheelchair propulsion. *Archives of Physical Medicine and Rehabilitation*, 77(9), 856-864.
- [19] Cooper, R. A., Boninger, M. L., Shimada, S. D., & Lawrence, B. M. (1999). Glenohumeral joint kinematics and kinetics for three coordinate system representations

- during wheelchair propulsion. *American Journal of Physical Medicine and Rehabilitation*, 78(5), 435-446.
- [20] Kulig, K., Rao, S. S., Mulroy, S. J., Newsam, C. J., Gronley, J. K., Bontrager, E. L., & Perry, J. (1998). Shoulder joint kinetics during the push phase of wheelchair propulsion. *Clinical Orthopaedics and Related Research*, (354), 132-143.
- [21] James, C. R. (1996). Effects of overuse injury proneness and task difficulty on joint kinetic variability during landing. In Stergiou, N. (2004). *Innovative analyses of human movement*. (pp.29-62) Champaign, Ill.; United States: Human Kinetics
- [22] Srinivasan, D., & Mathiassen, S. E. (2012). Motor variability in occupational health and performance. *Clinical Biomechanics*, 27(10), 979-993. doi: 10.1016/j.clinbiomech.2012.08.007
- [23] Hamill, J., Van Emmerik, R. E. A., Heiderscheit, B. C., & Li, L. (1999). A dynamical systems approach to lower extremity running injuries. *Clinical Biomechanics*, 14(5), 297-308.
- [24] Heiderscheit, B. C., Hamill, J., & van Emmerik, R. E. A. (2002). Variability of stride characteristics and joint coordination among individuals with unilateral patellofemoral pain. *Journal of Applied Biomechanics*, 18(2), 110-121.
- [25] Søndergaard, K. H. E., Olesen, C. G., Søndergaard, E. K., de Zee, M., & Madeleine, P. (2010). The variability and complexity of sitting postural control are associated with discomfort. *Journal of Biomechanics*, 43(10), 1997-2001.

- [26] van, d. H., Bruijn, S. M., Meijer, O. G., Hodges, P. W., & van Dieën, J. H. (2012). Mechanical coupling between transverse plane pelvis and thorax rotations during gait is higher in people with low back pain. *Journal of Biomechanics*, 45(2), 342-347.
- [27] Lomond, K. V., & Côté, J. N. (2010). Movement timing and reach to reach variability during a repetitive reaching task in persons with chronic neck/shoulder pain and healthy subjects. *Experimental Brain Research*, 206(3), 271-282. doi: 10.1007/s00221-010-2405-1
- [28] Madeleine, P., & Madsen, T. M. T. (2009). Changes in the amount and structure of motor variability during a deboning process are associated with work experience and neck–shoulder discomfort. *Applied Ergonomics*, 40(5), 887-894. doi: 10.1016/j.apergo.2008.12.006
- [29] Madeleine, P., Mathiassen, S. E., & Arendt-Nielsen, L. (2008). Changes in the degree of motor variability associated with experimental and chronic neck–shoulder pain during a standardised repetitive arm movement. *Experimental Brain Research*, 185(4), 689-698. doi: 10.1007/s00221-007-1199-2
- [30] Mathiassen, S. E., Möller, T., & Forsman, M. (2003). Variability in mechanical exposure within and between individuals performing a highly constrained industrial work task. *Ergonomics*, 46(8), 800-824.
- [31] Curtis, K. A., Roach, K. E., Brooks Applegate, E., Amar, T., Benbow, C. S., Genecco, T. D., & Gualano, J. (1995). Development of the wheelchair user's shoulder pain index (WUSPI). *Paraplegia*, 33(5), 290-293.

- [32] Hanavan Jr., E. P. (1964). a mathematical model of the human body. amrl-tr-64-102. AMRL-TR.Aerospace Medical Research Laboratories (6570th), 1-149.
- [33] Cooper, R. A., Boninger, M. L., Shimada, S. D., & Lawrence, B. M. (1999). Glenohumeral joint kinematics and kinetics for three coordinate system representations during wheelchair propulsion. *American Journal of Physical Medicine & Rehabilitation*, 78(5), 435-446.
- [34] Kwarciak, A. M., Sisto, S. A., Yarossi, M., Price, R., Komaroff, E., & Boninger, M. L. (2009). Redefining the manual wheelchair stroke cycle: Identification and impact of nonpropulsive pushrim contact. *Archives of Physical Medicine & Rehabilitation*, 90(1), 20-26.
- [35] Rice, I. M., Hsu, I. M. K., Jayaraman, C., Culp, S., Hsiao-Wecksler, E. T., Sosnoff, J. J., (2012, June). *Handrim Force Variability During Two Speeds of Wheelchair Propulsion*. Poster session presented at the Rehabilitation Engineering and Assistive Technology Society of North America, Baltimore, MD.
- [36] Rice, I. M., Pohlig, R. T., Gallagher, J. D., & Boninger, M. L. (2013). Handrim wheelchair propulsion training effect on overground propulsion using biomechanical real-time visual feedback. *Archives of Physical Medicine & Rehabilitation*, 94(2), 256-263.
- [37] de Groot, S., de Bruin, M., Noomen, S. P., & van, d. W. (2008). Mechanical efficiency and propulsion technique after 7 weeks of low-intensity wheelchair training. *Clinical Biomechanics*, 23(4), 434-441.

- [38] Newell, K. M., Deutsch, K. M., & Morrison, S. (2000). On learning to move randomly. *Journal of Motor Behavior*, 32(3), 314.
- [39] Reed, G. F., Lynn, F., & Meade, B. D. (2002). Use of coefficient of variation in assessing variability of quantitative assays. *Clinical and Diagnostic Laboratory Immunology*, 9(6), 1235-1239.
- [40] Sawatzky, B. J., Slobogean, G. P., Reilly, C. W., Chambers, C. T., & Hol, A. T. (2005). Prevalence of shoulder pain in adult- versus childhood-onset wheelchair users: A pilot study. *Journal of Rehabilitation Research and Development*, 42(3 SUPPL. 1), 1-8.
- [41] Hausdorff, J. M. (2009). Gait dynamics in Parkinson's disease: Common and distinct behavior among stride length, gait variability, and fractal-like scaling. *Chaos*, 19(2), 026113. doi: 10.1063/1.3147408